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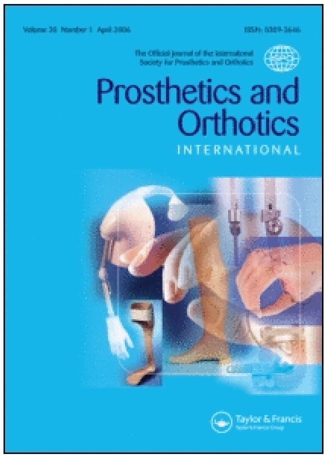

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Gait analysis in prosthetics: Opinions, ideas and conclusions

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Gait analysis in prosthetics: opinions, ideas and conclusions

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Abstract

A review was performed of the literature of the last eleven years (1990-2000) with the topic: "clinical use of instrumented gait analysis in patients wearing a prosthesis of the lower limb". To this end a literature search was performed in Embase, Medline and Recal. Forty-five (45) articles were identified for study from which 34 were reviewed.

The reviews were divided into five subtopics:
1) adaptive strategies in gait (12 studies);
2) the influence of different parts of the prosthesis on gait (12 studies);
3) pressure measurements in the socket in gait studies (4 studies);
4) the influence of the mass of the prostheses on gait (5 studies);
5) energy considerations in gait (2 studies).

A considerable part of the studies concerned the adaptive strategies of the amputee in walking and running and the evaluation of different prosthetic feet. All aspects and outcomes were reviewed concerning the clinical relevance.

Conclusion

Instrumented gait analysis in prosthetics provides better insights and knowledge of the different adaptive mechanisms of the body in walking with a prosthesis. Most instrumented gait studies are done in a gait laboratory. This is not comparable with the patient's home or work situation. Instrumented gait studies assess only the impairments. Assessment of disabilities and the subjective opinion of the patients' remain necessary.

It is the authors' opinion that the future of instrumented gait analysis in prosthetics is mainly for scientific research in which one should critically consider the number of patients to be studied, to show some clinically relevant differences. It is also important to test new prosthetic components, to investigate whether they improve the gait of the amputee wearing the prosthesis.

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Introduction

Since the introduction of instrumented gait analysis it has been used in measuring the gait of amputees walking with prostheses. To assess the value of this instrumented gait analysis in relation to clinical practice, the authors performed a literature search in Embase and Medline in which they searched for literature in the last eleven years (1990-2000) with the mesh headings "gait" and "amputee". Seventy-one (71) articles were found of which 26 were omitted, because instrumented gait analysis was not used. With Recal an additional 4 articles were found giving a total of 49 articles for further analysis. Studies included in the search had to meet the following criteria: 1) written in English language, 2) studied the use of instrumented gait analysis on trans-tibial amputees (TT), knee disarticulation amputees (KD) or trans-femoral amputees (TF), 3) were published in a journal. Finally only those studies which were, in the opinion of the authors, of clinical relevance were reviewed. Another 4 articles were excluded because they were abstracts, and 11 articles were excluded because of the lack of clinical relevance or because the main subject was other than gait analysis. In the end, 34 articles remained for reviewing. In all articles referenced the number of subjects, which were studied, are shown in brackets.

The articles are divided in five subgroups, which are all discussed separately. The subgroups are as follows:

1) adaptive strategies in gait;
2) the influence of different parts of the prosthesis on gait;
3) pressure measurements in the socket in gait studies;
4) the influence of the mass of the prostheses on gait;
5) energy considerations in gait.

Adaptive strategies in gait

This part refers to 12 studies. In 8 of these studies, the adaptive mechanisms of the amputee in walking and running were the main issues. One study concerned the velocity of walking comparing healthy volunteers and amputees (Hermodsson *et al.,* 1994). Two (2) studies concerned walking of children with an amputation (Engsberg *et al.,* 1992 (n=3); Engsberg *et al.,* 1993). One study investigated the influence of limb alignment on the gait (Yang *et al.,* 1991 (n=4)). In general the studies included only a few patients (n<10). In just two studies, more patients were involved (Hermodsson *et al.,* 1994 (n=24); Engsberg *et al.,* 1993 (n=22)). This of course influences the quality of the methodology of these studies.

Adaptive mechanisms in gait with a prosthesis

In recent years, several studies have been performed to assess adaptive strategies of patients wearing prostheses to attain a normal gait. In the TT-amputee, at the sound side, an almost normal electrical activity of the muscles was found (Czerniecki, 1996 (review); Buckley, 1999 (n=5)). However, the energy-absorbing function of the quadriceps muscles and of the ankle-foot unit was decreased at the amputated side (Czerniecki *et al.,* 1996 (n=5); Yang *et al.,* 1991). The reduction of the push-off force at the end of the stance-phase is partially compensated by the increased biomechanical work of the hip extensors (Czerniecki *et al.,* 1991). There is also an increased co-contraction activity of the knee muscles to get more stability or to act as an antagonist of the increased hip extension moment.

In the deceleration phase of the swing-phase at the sound side, there is an increased muscle work done by the hip and knee muscles that are responsible for the energy transmission from the limb to the body (Czerniecki, 1996). Through this mechanism the body receives forward acceleration at the time that the push off force of the ankle-foot unit is decreased. During running these adaptive mechanisms increase. There is no need for totally different muscle activation (Czerniecki and Gitter, 1992 (n=5); Buckley, 1999; Sanderson and Martin, 1996 (n=6)).

In the TF amputee, the sensory motor functions of the knee and ankle-foot are absent. This requires different adaptive mechanisms (Czerniecki *et al.,* 1996). During stance-phase the amputated side of the patient requires increased stability so flexion of the knee should be prevented. Knee-stability can be gained partially through the alignment of the prosthesis and by the choice of the knee-unit. Besides this adaptations in muscle activities also take place. In the first place, the patient does not allow his knee to flex at the amputated side in the first 30-40 percent of the stance-phase. He therefore uses his hip extensors in a closed kinetic circuit (Seroussi *et al.,* 1996 (n=16)). In the last part of the stance-phase (pre-swing) the hip flexors act in the same strength as normal despite the low weight of the prosthesis (30% of the weight of a normal lower limb). This is compensatory to the diminished push-off on the prosthetic side (Seroussi *et al.,* 1996).

The most important adaptation of the non-amputated side during stance-phase is the increased muscle work of the hip-extensors and the ankle-foot plantar-flexors to compensate for the decreased push-off of the prosthetic side (Seroussi *et al.,* 1996). Due to the increased plantar flexion work in mid-stance of the non-amputated side, the body centre of mass will rise, which allows the swing-phase of the prosthesis by raising the prosthesis off the ground (vaulting).

The study of Buckley (1999 (n=12)) cannot be generalised to a wider population because the study was on top athletes. During running of TF amputees there were large kinematic asymmetries between the contralateral limbs in contrast to the almost normal gait patterns in running of TT amputees (Buckley, 1999).

During swing-phase a modern knee unit can take over a part of the energy-absorbing functions in the early and late swing-phase of the quadriceps muscles and hamstrings respectively. In the literature no other adaptive mechanisms are described.

Hermodsson *et al.* (1994) described in their outstanding paper that the difference in gait performance between vascular (worse) and traumatic amputees (compared with healthy subjects) is not due to walking speed but due to the differences in active forces during push-off on both healthy and prosthetic legs. This could be an effect of the systemic disease in the case of the vascular amputees. In this respect, Pinzur *et al.* (1991 (n=12)) described that the difference found between active and limited walkers was the ability of active walkers to maintain quadriceps and hamstring muscle activity to compensate for their absent ankle activity.

Yang *et al.* (1991(n=4)) assessed the influence of limb alignment on the gait of TF amputees. An important factor was the angular displacement of the prosthetic thigh in gait, which changed to adapt to alignment changes. They also found significant effects of the alignment of the prosthetic foot and the socket on the ground reaction forces and joint moments at the prosthetic side. These data can be used to optimise an alignment or for example to change an alignment for increasing knee stability.

The influence of different parts of the prosthesis on gait

In total 12 studies were included in which 11 studies evaluated different prosthetic feet and one study analyses two knee units (Boonstra *et al.*, 1996 (n=28)). Again only studies with a low number of patients could be identified with one exception with a relatively large group of patients (Boonstra *et al.*, 1996). In 11 studies patients had a TT amputation and in one study a TF amputation. The studies concerning prosthetic-feet compare in most cases the characteristics in gait of the so-called conventional feet (i.e. SACH-foot) with the so-called energy storing feet (i.e. Flex-foot).

Influence of the prosthetic foot in gait

The different types of prosthetic feet influence, through their specific characteristics several kinetic and energy aspects in gait. This concerned mostly the joint motion (range of motion) and the energy absorbing and releasing aspects of the ankle-foot unit in stance phase. The range of motion of the ankle-foot unit can be described by the progressive plantar flexion of the foot in the early stance phase and the dorsiflexion of the foot in the late stance phase. Regarding to the energy aspects of the ankle-foot unit this concerns the energy absorption during heel strike, the storing of energy in the mid-stance phase but especially the release of energy to produce acceleration during walking.

Regarding these last aspects energy storing feet seem to have an advantage.

Range of motion

Several studies report an increased range of motion of energy storing feet in relation to conventional feet (Linden *et al.*, 1999 (n=2); Rao *et al.*, 1998 (n=9); Lehmann *et al.*, 1993 (n=10); Torburn *et al.*, 1994 (n=10)). This is dependent on which types of conventional or energy storing feet are compared. Most authors consider the SACH foot as the exemplar of the conventional foot and the Flex-foot or the Seattle foot as the representative of the energy storing foot as their opponents. However, conventional feet with a single axis, have a bigger range of motion (Postema *et al.*, 1997[a] (n=10)). One (1) study reports the early stance-phase in three different feet (Rao *et al.*, 1998). The conventional feet with a single axis showed a non-controlled plantar flexion mobility and the Seattle and Flex feet showed a decreased plantar flexion mobility in the early stance-phase (Rao *et al.*, 1998). Cortes *et al.* (1997 (n=8)) suggest that from a kinetic point of view the most determinant factor related to the behaviour of prosthetic feet is the presence or absence of a joint which allows for plantar flexion. These authors support the critera for the classification of prosthetic mechanisms as articulated and non-articulated (Cortes *et al.*, 1997).

Postema *et al.* (1997[a] (n=10)) suggest that dorsiflexion mobility of the ankle-foot unit during the late stance-phase influences the balance-control mechanism. An increase of dorsiflexion mobility during late stance phase gives an increase in the knee flexion torque, which can possibly decrease the stability factor at that level. However, a foot with a limited dorsiflexion in the late stance phase will give an extension moment to the knee, which will increase stability but also will give a limited "roll over" of the prosthesis.

For active amputees (walkers), a prosthetic foot with increased dorsiflexion mobility will be preferred and for less active amputees with balance problems a foot with less dorsiflexion mobility will be preferable.

Energy storing

The total energy behaviour of energy storing feet can be divided in three phases: 1) energy absorbing, 2) energy-storing and, 3) energy releasing aspects.

During heel strike energy absorption will take place through the hysteresis of the materials of which the feet are made. In principle, this means a loss of energy, when this absorbed energy will not be released at a later moment of the stance phase. In most conventional feet, the materials used will absorb this energy. Energy-storing feet will store the absorbed energy in their "spring-mechanism" and will partially release it at a later moment.

In analysis this is only indirectly measured, by means of ground reaction forces. The values of ground reaction forces are also dependent on many variables such as walking speed, cadence and body weight. Besides these, shoe characteristics also have great influence. Postema *et al.* (1997[a]) have carried out a study of the influence of the use of 4 different prosthetic feet (2 conventional and 2 energy storing feet) on gait analysis variables. In this study an analysis is described of the energy absorption (phase A1) and energy release (phase A2) during stance phase. The amount of energy absorption during heel strike is a measure of the stiffness of the hind foot. The lesser the absorption, the stiffer the foot. The biggest part of this absorbed energy will be lost in the foot materials itself. During late stance phase just before push-off, energy again will be stored in the foot (stiffness of the forefoot). A part of this stored energy will possibly be released during push-off. Postema *et al.* (1997[a]) found no differences in four feet concerning the degree of energy absorption (A1). There was a small, but significant difference in the energy release (A2) in favour of the two energy storing feet. The net difference between absorption and release of energy (measure of efficiency) was not significantly different. Other authors found greater differences.

During a walking speed of 1.5 m/s a factor of 2 to 3 was found as a measure for energy release during push-off comparing dynamic feet with the SACH foot (Gitter *et al.*, 1991 (n=5)). The same results were confirmed by Czerniecki *et al.* (1991) walking at 2.5 m/s. A comparable study done in TT amputated children, assessed also that the dynamic Flex-foot had a greater potential for reducing the energy cost of walking at 0.9 m/s (comfortable speed) and at 1.3 m/s (fast speed), than the SACH foot (Schneider *et al.*, 1993 (n=12)).

Since the energy absorbed at heel-strike will not be given to the forefoot, great absorption will give a considerable loss of energy. However, a lot of patients will consider this as very comfortable (Postema *et al.*, 1997). These were important findings for the individual patients, but because of the small number of patients it is not possible to generalise these findings. It can be used as empirical knowledge and not as evidence based knowledge.

In a study by Postema *et al.* (1997[b]) in which the patients were asked their opinion about these feet, this feeling about comfort was recorded. Besides energy absorption also other aspects are considered to be important such as: a "springing" push-off, flexibility, stability etc. Other authors also came, indirectly, to this conclusion as they found that an increased dorsiflexion of for instance the Flex-foot, will ease the roll over (Linden *et al.*, 1999; Rao *et al.*, 1998; Lehmann *et al.*, 1993). This characteristic is not as much related to the energy storing principle of the foot but more to the ankle mechanism of the foot.

Lehmann *et al.* (1993 (n=9)) did a very interesting study of the moment of energy release during stance-phase. They calculated the time between the moment of mid-stance (energy absorption) and the moment of push-off (energy release). With the help of the calculated foot-specific frequency of energy absorption and release by a load from a mass of 68 kg (i.e. the leafspring principle in the energy storing foot), they concluded that the moment of energy release always comes before push-off starts and therefore does not coincide with push-off. In the same study, Lehman *et al.* (1993) also concluded that there is a clear difference between the Flex-foot and the SACH foot on the landing-phase of the sound limb. This loading response on the sound side and thus the load transfer from the prosthetic to the sound side was lower when using the Flex-foot than when using the SACH foot. He hypothesised that the absence of the push-off activity of the SACH foot was responsible for this phenomenon.

Powers *et al.* (1994 (n=10)) also concluded that the SACH foot gives the highest and the Flex-foot the lowest loading response at heelstrike to the sound limb. Contrary to Lehman they hypothesised that the reason was that the greater dorsiflexion possibility of the Flex-foot during late stance phase meant that the body centre of gravity has to be less elevated resulting in a lower loading response at heel strike.

Knee components

Only one study compares two different knees using gait analysis (Boonstra *et al.*, 1996). These authors compare a conventional 4 axis knee of Otto Bock (3R20) with the pneumatic controlled swing phase knee (Teh Lin). The authors also combined objective gait analysis with a subjective judgement of the amputee. The knee with a pneumatic swing phase control allows the amputee to walk at higher speed; the comfortable speed stays the same for both knees. Walking with the Teh Lin knee showed more symmetry. The range of motion of the Teh Lin knee during swing phase was smaller than the range of motion of the 3R20. The subjective opinion of the amputees regarding the comfort with the higher walking speed was in favour of the knee with the pneumatic swing control. This last study was not of a double blind design so the research conclusions should be regarded as having limitations.

Pressure measurements in the socket in gait studies

Four (4) studies involve this topic. Two (2) of those concern a case study, in which the pressure distribution in the socket is measured during gait in a handcast fabricated patellar-tendon-bearing PTB socket (Convery and Buis, 1998 (n=1)) and in a hydrocast-method fabricated socket (Convery and Buis, 1999 (n=1)). In both sockets there was a relationship between the pressure distribution in the socket and the projection of the ground reaction force. During heel strike there was a higher-pressure proximal-posterior in the socket and after mid-stance especially proximal-anterior. With the hydrocast method constructed socket, the pressures were lower and more balanced. Both studies are performed correctly and thoroughly but the results cannot be generalised, only being considered as an indicator.

Two (2) studies analysed the relation between alignment changes and pressure distributions in a PTB socket during gait (Sanders *et al.*, 1998 (n=2); Sanders *et al.*, 2000 (n=3)). In these studies the variance between two different measurements in time (one week) was greater than the variance in pressure distribution due to alignment changes. The same criticism as mentioned earlier can be used here as these were case series with only 2 and 3 subjects. In the study of Sanders *et al.* (2000), the authors also conclude that effective techniques to accommodate time related stump volume fluctuations have a greater impact on maintaining consistent interface stress distributions than do alterations in cadence or componentry. Pressure measurement research in prosthetics seems to be a big challenge for future researchers.

The influence of the mass of the prostheses on gait

This topic is covered by 5 studies. Two (2) studies concern patients with a TT amputation, and 2 other studies concern patients with a TF amputation. In all these 4 studies, weight is added to the prostheses in order to study the changes in gait. The fifth study is a review article (Selles *et al.*, 1999).

In the study of Gailey *et al.* (1997 (n=10)) a mass of 454 or 907 grams was added to a trans-tibial prosthesis. The added mass was evenly distributed over the length of the prosthesis shank. No significant difference in energy costs was found. The energy cost of the TT amputees was about 13% more than in healthy subjects. It must be taken into account that the 10 young subjects (37.8 years; SD=10.4) assessed in this study were 8 traumatic amputees, 1 congenital amputee and 1 patient amputated because of tumour. This cannot be considered as a normal amputee group as the majority of amputees suffer from vascular disease.

Hillery *et al.* (1997 (n=1)) added up to 1460 grams to the trans-tibial prosthesis and they measured equal ground reaction forces. The added mass however changed the kinematics. The cadence was the highest in the prosthesis with the lowest mass. The extension and flexion of the hips at both sides increased slightly when mass was added to the prosthesis. In the kinematics of the knee no great changes were found.

Hale (1990 (n=6)) concluded that added mass to a prosthesis increases the hip eccentric muscular effort (especially the hip extensors) during the deceleration of the limb during the end of the swing phase. During acceleration there was no increase in muscular effort. Gitter *et al.* (1997 (n=8)) studied the influence of adding mass to a prosthesis and the mechanical work required for limb movement during swing and came to another conclusion. When 1340g were added to the prosthesis, there was an increased concentric muscle effort needed for the hip flexors in order to accelerate the prosthesis into the swing phase. This increased acceleration work was compensated by a comparable increase in recovery of leg energy during terminal swing deceleration. They concluded that the amputee appears to minimise the adverse effect of added mass to a prosthesis by effectively conserving the additional mechanical work needed for the propulsion of the limb.

In the systematic review of Selles *et al.* (1999), the authors concluded that within the range of masses studied, kinematics and energy aspects of prosthetic gait do not change.

Energy considerations in gait

Energetic aspects are to some extent already described above. Another 5 studies were found of which the two largest are described.

Gailey *et al.* (1994 (n=39)) compares 39 TT amputees with 21 non-amputees. The amputees used 16% more energy and walked 11% more slowly than the non-amputees. No significant correlation was found between energy cost of ambulation and the stump length or mass of the prosthesis. With stratification for a long or short stump length, there was a slight but significant difference in energy costs in favour of the long stump. Points of criticisms are that the amputee group and the control group differ in age and that the amputee group is relatively young (mean is 47 years; SD=16). Comparing with the general prosthesis users it can be debated whether the results could be generalised to the wider populations of amputees.

Jaegers *et al.* (1993 (n=11)) analysed the efficacy of gait in 11 TF amputees and compared this with 6 non-amputees. In the non-amputee group the comfortable walking velocity is also the most metabolic efficient velocity. In TF amputees the comfortable walking velocity is lower than the most metabolic efficient walking velocity. However, if both, non-amputees and amputees walk at their most efficient velocity, no difference in energy expenditure in units of time could be detected. In this study however the number of amputees was small and taking into account that all amputees had different components in their prostheses, no general conclusions could be drawn.

Discussion

Gait analysis in prosthetics allows better insights and knowledge in the different adaptation strategies of the human body in order to walk as normally as possible with a prosthesis. Therefore these insights can give us tools to develop better gait training programmes for amputees and give also the knowledge for development of new components. It must be realised that the results of these studies cannot be generalised to every amputee. The individual amputee wearing a prosthesis may have developed a personal adaptation. Furthermore almost all subjects assessed in the studies were without co-morbidity in contrast to the most commonly encountered vascular amputees. Finally, these are questions about the methodology in many papers because of the small number of subjects assessed.

Medline, Embase and Recal were searched for publications about instrumented gait analysis. There is a great overlap between Medline and Embase. Using Recal, an additional 4 articles were found. There is a possibility with these three databases that there is a restriction on the number of studies recognised by the authors.

If all studies are excluded with less than 10 patients (arbitrarily chosen), and/or inadequate methodology just a few of studies could be included. For instance, only 14 papers concerned studies with 10 or more patients, of which 2 (Postema, and 1997[a,b]) assessed the same patients. Conclusions of most studies, therefore, are not evidence based and there is limited scientific value. The authors did not perform a systematic review because it was expected, as Selles *et al.* (1999) also concluded, that most gait studies are of moderate methodological quality. Nevertheless, discussions and consideration of many studies gives tools to consider different aspects of the clinical use of instrumented gait analysis in patients with a prosthesis.

Most data are gathered from instrumented gait studies performed in gait laboratories. This is not comparable with the patients' home or work situations. Instrumented gait studies assess only the impairments. This is a good starting point but assessment of disabilities and the subjective opinion of the patients remain necessary. For example; stability of the prosthetic foot is partly a subjective perception.

Postema *et al.* (1997[b]) stated that the subjective perception of stability correlated strongly with the kind of activities of daily living. Two (2) of the 10 patients in their study worked in the construction industry. They experienced a flexible foot as more stable. On the contrary, patients who walked mostly on flat floors found a stiff foot more stable. This subjective opinion is not part of the assessment by instrumented gait analysis.

Instrumented gait analysis is regarded as a specific measurement, mostly by fixed parameters such as velocity, duration of stance and swing-phase, floor reaction forces etc. A medical doctor or prosthetist or bioengineer or physiotherapist however bases the clinical implications on the interpretation of gait characteristics. The different conclusions in the literature regarding conventional and energy storing feet may be a result of this.

The future of instrumented gait analysis in prosthetics will be mainly for scientific research in which one should critically consider the number of patients, to be studied, to show some clinically relevant differences. It will be increasingly important to test new prosthetic components, to investigate whether these new components really influence, in a good way, the gait of the amputee wearing a prosthesis. Instrumented gait analysis for clinical use of individual patients is of interest but should be regarded critically.

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